A Miniature "Scintillator-Fiberoptic-PMT" Detector System for the Dosimetry of Small Fields in Stereotactic Radiosurgery

A. S. Beddar, T. J. Kinsella, A. Ikhlef, and C. H. Sibata

***Abstract*—Linear accelerator-based stereotactic radiosurgery is an important modality in the management of localized intracranial abnormalities and often is the only option for treatment of certain lesions. The quantification of the absorbed radiation doses delivered for this modality has been a challenge due to the small size of the treatment fields (typically 5–30 mm collimated fields) and the type of detectors presently available. This paper presents comparative dose measurements performed using a miniature (1.6×10^{-3} cm^3) plastic scintillator detector to the commonly used detectors for the dosimetry of small photon fields used for stereotactic radiosurgery. The spatial resolution of the plastic scintillator detector was compared to that of a p-type Si diode and a 0.1-cm^3 ionization chamber. Small field dosimetry parameters (beam profiles, percent depth doses, and dose output factors) using the scintillation detector, the 0.1-cm^3 ionization chamber, the p-type Si diode, and radiographic film are presented. This comparative study shows that designing a miniature scintillator detector system is feasible and when compared to other benchmark detectors is an appropriate detector for the dosimetry of small stereotactic radiosurgery photon fields.**

***Index Terms*—Dosimetry, optical fibers, photodetector, scintillator, stereotactic radiosurgery.**

I. INTRODUCTION

STEREOTACTIC radiosurgery is an important treatment modality in the management of localized intracranial abnormalities and often is the only option for treatment of certain lesions. The quantification of the absorbed radiation doses delivered for this modality has been a challenge due to the small size of the treatment fields (typically 5–30 mm diameter fields) and the type of detectors presently available [1]–[3]. The current practice for dosimetry in stereotactic radiosurgery involves the use of ionization chambers, diodes, thermoluminescent detectors (TLDs), and film dosimetry [4]. However, there are problems involved in their use. Due to the lack of charged particle equilibrium (CPE), knowledge of the spectrum at all points is required to correct the dose measurements. Additional problems for ionization chambers and diodes include dose perturbation of primary and charged particle fluences caused by replacement of the medium by the detector at the measurement point. The development and the design of a unique miniature "scintillator-Fiberoptic-PMT" system, based on previous work [5], [6], has enabled us to successfully resolve these problems and measure doses accurately with a high spatial resolution.

Plastic scintillators have been used extensively in high-energy nuclear particle physics and particularly in spectroscopy experiments in the past. Such detectors had fairly large-sized scintillators and were mainly applied for beam time structure, energy of events, and time of flight measurements. Recently, plastic scintillator detectors have been successfully applied to high-energy photon and electron beam dosimetry, field mapping, and x-ray imaging [5]–[10]. Some of the most attractive advantages of using these detectors are:

1) that the usual conversion of absorbed dose from one medium to another can be avoided because the scintillator material is water-equivalent;
2) they cause minimal perturbations of the radiation field being measured;
3) their small size permits accurate dose measurements in regions of high dose gradients;
4) if made small enough, the presence or absence of charged particle equilibrium becomes irrelevant [3], [5], [6], [9].

Pain *et al.* have just recently used scintillating plastic fiber to quantitatively measure neuropharmacological radiotracer kinetics and dose distribution in awake and unrestrained animals. A small intracerebral probe, surgically implanted in an animal brain, was used to measure the neurophysiological activity in awake and unrestrained small animals [11].

Another category of detectors, namely, diamond detectors, has been used successfully by Rutsgi and Frye [12] in 1995 and by Heydarian *et al.* [13] in 1996, but their application is not widespread. This is due to the poor availability of the detector and its high price. In 1995, as mentioned by Rutsgi and Frye, the waiting period for a diamond detector was approximately 12 months, and it was approximately seven to eight times more expensive than an ionization chamber. Unfortunately, these diamond detectors of small sensitive volume, which have shown promise in the dosimetry of stereotactic radiosurgery fields due to their interesting characteristics, are not currently available.

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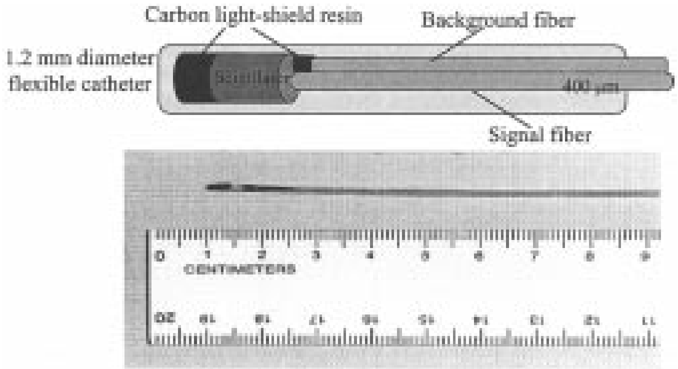

Fig. 1. A schematic diagram and a scaled photograph of the proximal end of the Scintillator-Fiberoptic-PMT detector system.

II. Description of Scintillator-Fiberoptic-PMT Detector System

The main basic component of this miniature detector is based on a similar design that has been previously described [5]. The detector system consists of a miniature plastic scintillator (BC 400, 1 mm diameter by 2 mm long) optically coupled to a single 400-μm pure silica radiation resistant fiber[1] inserted inside a 1.2-mm outer diameter flexible catheter. A shielded second identical fiber (background) running in parallel to the signal fiber is used to account for the radiation-induced light in the fibers. The background signal is subtracted to yield only the signal from the scintillator. Each single fiber was optically coupled to an independent 10-mm head-on-type photomultiplier tube manufactured by Hamamatsu.[2] A schematic diagram and a scaled photograph of the proximal end of the detector are shown in Fig. 1. The design of this miniature scintillator detector is similar to the design reported by Beddar *et al.*in 1992. The major differences are 1) a smaller scintillator size (the length was reduced from 4.0 to 2.0 mm) and 2) both the signal and background light guides are each composed of a single fiber instead of seven hexagonally packed fibers. The present 400-μm core diameter fiber has a diameter equal to 440 μm, including the cladding, and a diameter equal to 480 μm including the cladding and the external buffer. The fibers used previously had a core diameter equal to 200 μm, a diameter equal to 220 μm including the cladding, and a diameter equal to 240 including both the cladding and the external buffer. By reducing the scintillator size by one-half, we were still able to achieve a good mechanical and optical coupling between the scintillator and the fiber.

The background noise in this system arises from two sources. The first source is the inherent dark current of the PMT, which is approximately three-to-four orders of magnitude smaller than scintillator signals generated for typical radiotherapy dose rates (2–4 Gy/min), and was therefore considered negligible. The second source arises from the radiation-induced light inside the fiber itself. We have shown that the induced light for pure silica fibers is almost entirely due to Cerenkov radiation [14]. If fibers other than pure fused silica or high-purity glass fibers are used, then the radiation-induced light would be a combination of Cerenkov, luminescence, and probably some fluorescence [6], [14], [15]. In such a case, the combination of these background sources could interfere with the scintillation light. In our case, due to the small fiber length being irradiated (approximately half of the cone diameter) and because we used pure silica fibers, the signal-to-noise ratio was fairly large; therefore, Cerenkov was neglected.

[1]Polymicro Technologies, "Low Loss Bundle Fiber," Phoenix, AZ.

[2]Hamamatsu, Corporation, "Photomultiplier Tubes," Bridgewater, NJ.

III. Materials and Methods

Stereotactic radiosurgery techniques utilized to treat small brain lesions are essentially performed using a combination of multiple isocentric arc irradiations with small fields centered in the tumor volume target [16]. The word "radiosurgery" refers to the ability to perform brain surgery using radiation instead of a scalpel and the ability to deliver high radiation doses to small intracranial lesions in one single treatment. A medical linear accelerator (linac) was chosen as the source of ionizing radiation and provides a 6-MV photon beam. A linac can, while rotating about a point called isocenter, deliver a high dose of photon radiation to a specified tumor target, but the dose distribution outside the target is widely distributed due to the large penumbra of the conventional collimators of a linac. These collimators are inadequate for stereotactic radiosurgery because the dose delivered outside of the target volume is unacceptable and usually exceeds normal tissue dose tolerance. Therefore, these techniques use a specially designed collimator, which accommodates cylindrical cone inserts to improve the alignment and minimize the penumbra of the beam, thereby accentuating the rapid dose falloff outside the lesion. Such inserts are made from cerrobend (lead, tin, and bismuth alloy), are 10 cm thick and are available with diameters of aperture varying from 5 to 30 mm in steps, typically of 5 mm [16]–[18]. In this work, dosimetric data for the 5-, 10-, 20-, and 30-mm stereotactic cones (collimator-and-cone inserts) are presented using the miniature plastic scintillator detector as described above and compared to the most commonly used radiation detectors. Fig. 2 shows these detectors (the plastic scintillation detector, a 0.1-cm^3 PTW ionization chamber and an Si photon diode) side by side. The carbon coating covering the plastic catheter of the scintillator detector has been removed by dissolving it with an acetone solution for illustrative purposes.

The scintillator-fiberoptic-PMT detector system was first tested using the 6-MV photon beam using a 10 $\times$ 10 cm^2 radiation field size at a nominal source-to-surface distance (SSD) equal to 100 cm to determine the general detector's characteristics and compare them to those of original detector system [6]. The detector was fixed rigidly to a water-equivalent holder and placed in a water phantom scanning system. A computer-controlled water phantom scanning system was used to vary the depth of the detector along the central axis of the radiation beam. The properties of the detector were studied at the depth of maximum dose buildup ($d_{\max}$), which for a 6-MV photon beam is equal to 1.5 cm. The surface of the water was set to 100 cm SSD. For the stereotactic radiation field measurements, the same water phantom scanning system was used with an SSD of 100 cm to the water surface.

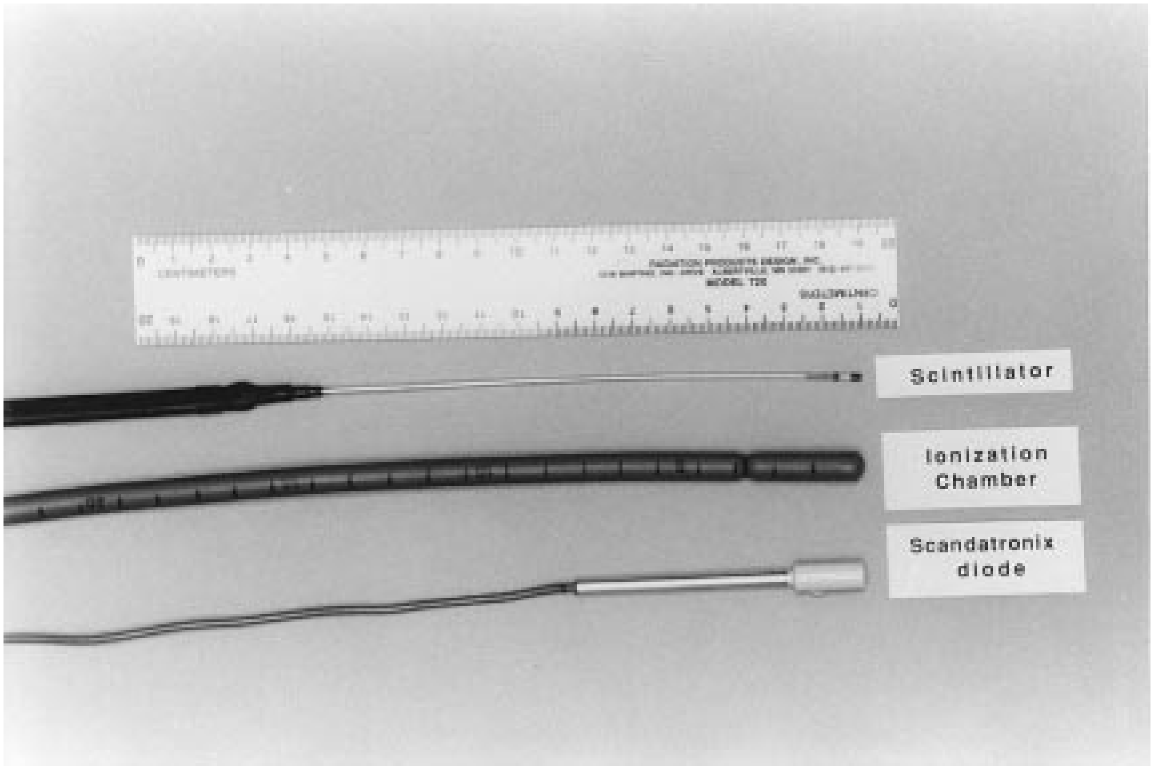

Fig. 2. The scintillator detector, a 0.1-cm^3 PTW ionization chamber, and an Si photon diode side by side to scale.

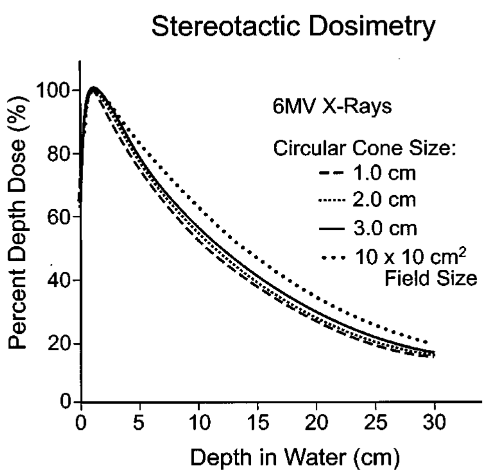

Fig. 3. Absorbed dose as a function of depth in water, normalized to depth $d_{\max}$ for the 10-, 20-, and 30-mm cone using the scintillator detector. The percent depth dose for the 10 $\times$ 10 cm^2 reference field size is also shown.

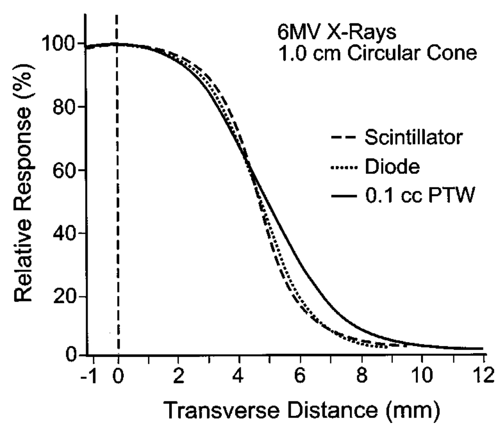

Fig. 4. Beam profile measurements made with the scintillator detector, the diode, and the 0.1-cm^3 PTW ionization chamber for the 10-mm stereotactic cone.

IV. Results

A. Characteristics of the Scintillator-Fiberoptic-PMT Detector System

The detector stability, reproducibility, and linearity were found to be similar to the first prototype detector, as described by Beddar *et al.* [6]. The detector response was found to be linear with absorbed dose and independent of dose rate, as expected and previously reported [6], [9]. To verify the system integrity and validate the use of this miniature scintillator system, dose measurements as a function of depth in water were performed on a 10 $\times$ 10 cm^2 field (the reference field size for beam calibrations) using both the plastic scintillator detector and a 0.6 cm^3 Farmer ionization chamber with a dose calibration traceable to National Institute of Standards and Technology calibration laboratory. Excellent agreement was obtained between both detectors (within less than 0.5% at all depths). The resulting dose measurement for the 10 $\times$ 10 cm^2 field shown in Fig. 3 corresponds to the curve obtained with the scintillator detector.

B. Relative Depth Dose Measurements

Dose measurements as a function of depth in water using this scintillation detector were made on small stereotactic radiosurgery beams with diameters of 5, 10, 20, and 30 mm. Good

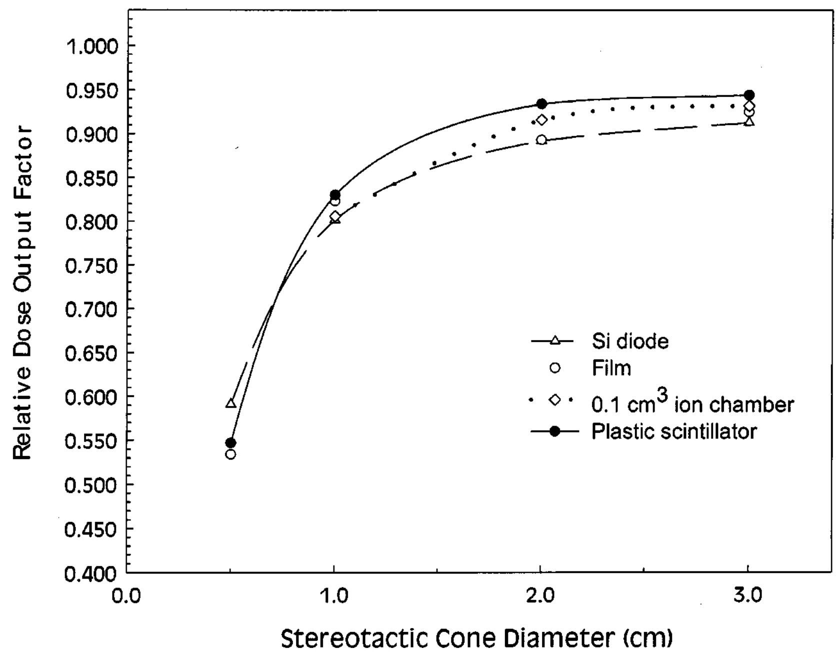

Fig. 5. Dose output factors for the different stereotactic collimator cones using the Si photon diode, radiographic film, the 0.1-cm^3 PTW ionization chamber, and the plastic scintillator detector.

agreement was found between the scintillator and the ionization chamber down to a field diameter of 20 mm. Below 20 mm, the ionization chamber did not have sufficient spatial resolution to provide an accurate measure of the dose. In addition, the expected shift of the depth of maximum dose ($d_{\max}$) toward the surface with decreasing field diameter that is seen with the scintillator is not seen at all with the ionization chamber and only slightly with the diode. Fig. 3 shows the absorbed dose normalized to depth $d_{\max}$ for the 10-, 20-, and 30-mm cones using the scintillation detector. The percent depth dose for the 5-mm cone is almost identical to the 10-mm cone size and is not shown in Fig. 3.

C. Spatial Resolution of the Scintillator Detector

Beam profile measurements made with the scintillator detector, the Si diode, and the 0.1-cm^3 PTW ionization chamber were acquired for field diameters of 30, 20, and 10 mm. The results show that the scintillation detector exhibits a higher spatial resolution than the 0.1-cm^3 PTW ionization and the Si diode. This is best illustrated for the 10-mm diameter beam as shown in Fig. 4. The scintillator detector shows a sharp falloff (i.e., high spatial resolution) inside the beam and outside the beam but starts losing this resolution at about 2 mm outside the beam edge (at a transverse distance of approximately 7 mm). This is due to a detector signal-to-noise reduction. In this case, the scintillator signal approaches a zero value since the detector is outside of the radiation beam. Therefore, the signal becomes comparable to the noise (i.e., dark current of the PMT).

D. Collimator Cone Output Factors

The dose output factors along the central axis of the radiosurgery beam for different stereotactic collimator cones, relative to the dose output of a 10 $\times$ 10 cm^2 field, are shown in

TABLE I
RELATIVE OUTPUT FACTORS FOR DIFFERENT STEREOTACTIC CONES USING AN SI-PHOTON DIODE, RADIOGRAPHIC FILM, 0.1-cm^3 IONIZATION CHAMBER, AND PLASTIC SCINTILLATOR

Cone Size	Diode	Film	Ion Chamber	Scintillator
0.5	0.591	0.534	*****	0.547
1.0	0.802	0.823	0.806	0.830
2.0	0.892	0.893	0.916	0.934
3.0	0.913	0.925	0.932	0.944

Table I. Note that the output for the 5-mm cone cannot be measured using an ionization chamber because of its large size. The relative output factors are normalized to the 10 $\times$ 10 cm^2 square field. The dose output factor for the 10 $\times$ 10 cm^2 square field is chosen to be equal to 1.000, and the dose calibration for this field size (the reference field size of the Linac) is 1 cGy per monitor unit delivered by the linear accelerator at ($d_{\max}$). The plastic scintillator, because of its spatial resolution and its miniature detecting volume size, provides the most accurate and reliable measurement of dose. The other detectors, due to their larger detecting volume size, integrate dose over a much larger volume for which the edges are close to the penumbra where the dose starts to fall off. Fig. 5 illustrates these results, showing that the largest relative dose output factor discrepancy is about 5% (for the 20 mm between the scintillator and either the film or the diode). The agreement between the small ionization chamber and the scintillator is within less than 2%.

V. Conclusion

This comparative dosimetry study shows that this scintillation detector is a suitable detector for dosimetry of small photon fields such as those encountered in stereotactic radiosurgery. This detector system allows for greater confidence in the determination of absorbed dose distributions and the determination of stereotactic collimator output factors when compared to the other benchmark detectors. The properties of the scintillation detector, including tissue equivalence, energy independence, dose linearity, high spatial resolution, and real-time measurement, make this detector an ideal choice for dose mapping measurements without the need to apply corrections to measurements needed by other detectors (film, Si diode).

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